Prediction of the Benefit of Motion-Compensated Reconstruction for Whole-Heart Coronary MRI

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INTRODUCTION – Respiratory motion represents a major challenge in free-breathing whole-heart coronary MR angiograpy (CMRA). Several approaches have been proposed to compensate for the effects of respiratory motion and to recover diagnostic image quality. Weighted iterative reconstruction[1,2], which is equivalent to retrospective soft-gating, aims to reconstruct a consistent sub-set of the acquired data. However, the soft-gating may lead to artifacts due to increased sub-sampling. Motion-compensated[3,4] (MoCo) reconstruction techniques promise to overcome this by incorporating all acquired data during image reconstruction using a motion model. Unfortunately, this comes at the price of longer processing times, which may not be justified in terms of the resulting signal-to-noise ratio (SNR) improvement in all cases. This work proposes a method to predict the benefit of MoCo over weighted reconstruction directly after data acquisition. In-vivo experiments were performed in 15 healthy volunteers and both methods were compared in terms of SNR.

MATERIALS and METHODS – After the acquisition, readouts can be binned according to the respiratory phase d they were acquired in, which can be determined e.g. from a navigator[5] or using the self-navigation principle[6]. A respiratory phase histogram $h(d)$ characterizes the respiratory distribution of the acquired data. Two such histograms are shown in Fig. 1, the first with a more uniform distribution than the second. For weighted reconstruction, let $w(d-d')$ be a function that assigns a weight to each respiratory phase d, where d' is the reference phase being reconstructed. Phases more distant to the reference phase receive a lower weight, e.g. with a box or Gaussian function centered at d'.

Then, the premise for the prediction of the SNR improvement of MoCo over weighted reconstruction is that acquisitions with a more uniform histogram will benefit from MoCo reconstruction, because more data would be excluded in the weighted reconstruction. It is based on a feature f_i for the i^{th} test subject, which is computed as the ratio of the average available data "within" the weighting function, i.e. where $w(d-d')>0$, over the average ratio outside, $f_i = \left(\#P_i^{-1}\Sigma_{d\in P_i} h_i(d)\right)/\left(\#\bar{P}_i^{-1}\Sigma_{d\in \bar{P}_i} h_i(d)\right)$, where # denotes the cardinality of a set, $P_i = \{d \mid w_i(d-d')>0\}$ is the set of respiratory phase indices having a positive weight, $w_i(d)$ is the weighting kernel and $h_i(d)$ the histogram of test subject i. Finally, the SNR improvement r_i is predicted as $r_i \approx a\cdot\exp(b\cdot f_i)+c$ from a regression with coefficients a, $b<0$ and c to be determined from training data. The exponential model was chosen because theory suggests a decreasing function and observed values show non-linear behavior.

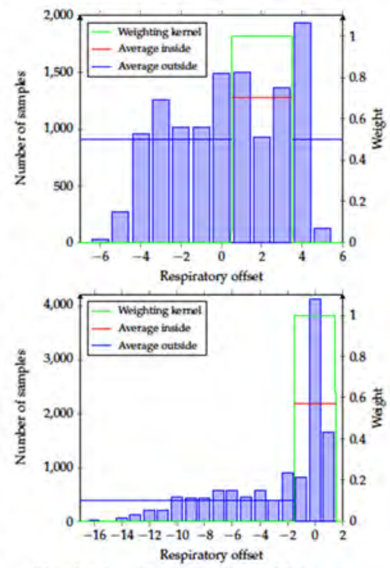

Fig. 1. Respiratory phase histograms with a near-uniform (top) and non-uniform (bottom) distribution.

In-vivo experiments were performed in 15 healthy volunteers (male, age 25 ± 5) on a 1.5 T clinical MR scanner (MAGNETOM Aera, Siemens AG, Healthcare, Erlangen, Germany). Signal reception was performed using the body and spine matrix coils. The parameters of the 3-D free-breathing, ECG-triggered, T2-prepared, fat-saturated, volume-selective bSSFP prototype sequence were: TR/TE 4.0/2.0 ms, flip angle 90°, receiver bandwidth of 849 Hz/Px, acquired matrix 256x218x150 that was interpolated to 256x256x176 in reconstruction for an isotropic voxel size of 1.05 mm^3. An incoherent sampling of the Cartesian phase-encoding plane was achieved with the spiral phyllotaxis pattern[7]. At the beginning of each heartbeat, an additional readout for the detection of respiratory motion was acquired. The sampling pattern was generated for 398 heartbeats containing 30 readouts. The SNR was computed for both types of reconstruction, followed by a leave-one-out cross-validation of the regression to determine the correlation coefficient and root-mean-square error (RMSE).

RESULTS and DISCUSSION – The average reconstruction time was 2 min for weighted and 37 min for MoCo reconstruction. The images with the highest and lowest feature value are shown in Fig. 2 and correspond to the highest and lowest SNR difference. A prediction of the SNR increase from the respiratory phase histogram could be made with a correlation coefficient of 0.77 and RMSE of 1.7 %. The fitted regression coefficients for the used weighting function were $a=0.52$, $b=-1.31$ and $c=0.12$. The histogram-based feature is a good indicator for the expected SNR benefit, as seen in Fig. 2. From the limited training data set of 15 volumes, the SNR benefit can already be predicted directly from the respiratory phase histogram with reasonable accuracy to support a choice for the reconstruction method to be used. Above a predefined threshold, MoCo reconstruction can be applied with the reasonable expectation that it will improve image quality. We expect that this prediction could be improved with additional training data for the regression.

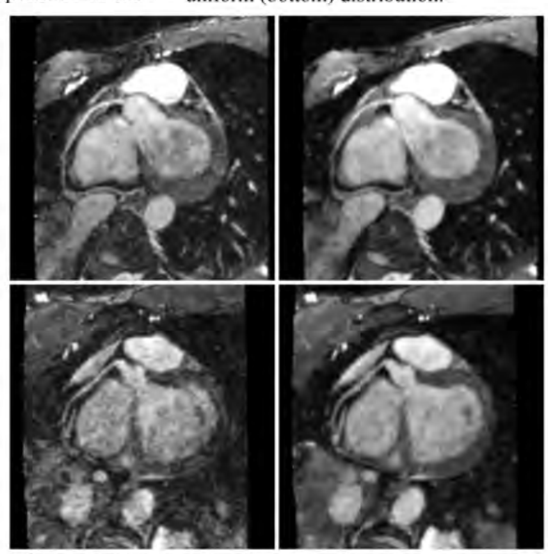

Fig. 2. Weighted (left column) and motion-compensated (right column) reconstructions for the volunteers with the lowest (top row) and highest (bottom row) feature value f_i.

CONCLUSIONS – We have shown that the benefit of MoCo reconstruction over the faster weighted reconstruction in terms of SNR can be predicted directly from the acquired data to allow a decision for one approach or the other. In cases where MoCo is unnecessary, the reconstruction is almost 20 times faster. We expect that these insights are transferrable to other applications of motion-compensated reconstruction outside of coronary imaging. It will have to be determined if a retraining will be necessary if the acquisition protocol changes, as it was fixed in our experiments. A validation on patient data is desirable.

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